DECLARATION

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Dated this 21st day of June, 2005

2000-324904

[Name of Document] SPECIFICATION

[Title of the Invention] NOISE SUPPRESSING APPARATUS AND STORAGE MEDIUM

[Scope of Demand for Patent]

[Claim 1] A noise suppressing apparatus for suppressing noise components included in a radiographic image, comprising:

smoothing means for processing an input image signal which represents an input radiographic image by using a smoothing filter such that the input image signal is smoothed; and

noise properties calculating means for calculating noise properties of the input image signal based on information indicating an exposure dose with which the radiographic image has been produced,

wherein the smoothing means adapts a characteristic of the smoothing filter which is used for performing the smoothing, based on the calculated noise properties.

[Claim 2] The noise suppressing apparatus as defined in claim 1, further comprising a band-limited-image-signal generation means for unit for generating a plurality of band-limited image signals respectively representing a plurality of band-limited images belonging to a plurality of different frequency bands, based on the input image signal,

wherein the smoothing means processes the plurality of band-limited image signals by using the smoothing filter so as to smooth each of the plurality of band-limited image signals.

[Claim 3] The noise suppressing apparatus as defined in claim 2, wherein the band-limited-image-signal generation means generates the plurality of band-limited image signals by performing multiresolution decomposition on the input image signal.

[Claim 4] The noise suppressing apparatus as defined in claim 2 or 3, wherein the noise properties calculating means obtains the noise properties also based on information locally calculated from pixel values in the vicinity of a pixel of

interest in at least one of the plurality of band-limited image signals.

[Claim 5] The noise suppressing apparatus as defined in claim 4, wherein the noise properties calculation means obtains a pixel vector at the pixel of interest in a band-limited image represented by at least one of the band-limited image signals, and detects an orientation of an edge as the noise properties using the pixel vector as the locally calculated information, and

the smoothing means adapts a characteristic of the smoothing filter so that the smoothing is performed along the orientation of the detected edge.

[Claim 6] The noise suppressing apparatus as defined any one of claims 1 to 5, wherein the smoothing means comprises a plurality of filters for smoothing a radiographic image in predetermined directions to respectively different degrees, and the smoothing means adapts a characteristic of the smoothing filter by selecting one of the plurality of filters based on the calculated noise properties.

[Claim 7] A computer-readable storage medium storing a program which instructs a computer to execute on an input image signal representing an input radiographic image a process for suppressing noise components included in the radiographic image,

wherein the program instructs the computer to carry out the steps of:

calculating noise properties of the input image signal based on information indicating an exposure dose with which the radiographic image has been produced;

adapting based on the calculated noise properties a characteristic of a smoothing filter used for subjecting the input image signal to smoothing; and

processing the input image signal by using the adapted smoothing filter such that the input image signal is smoothed.

[Claim 8] A computer-readable storage medium storing a

program which instructs a computer to execute on an input image signal representing an input radiographic image a process for suppressing noise components included in the radiographic image,

wherein the program instructs the computer to carry out the steps of:

generating a plurality of band-limited image signals respectively representing a plurality of band-limited images belonging to a plurality of different frequency bands, based on the input image signal;

calculating noise properties of the input image signal based on information indicating an exposure dose with which the radiographic image has been produced;

adapting based on the calculated noise properties a characteristic of a smoothing filter used for processing the plurality of band-limited image signals such that each of the band-limited image signals is smoothed; and

processing the plurality of band-limited image signals by using the adapted smoothing filter such that each of the band-limited image signals is smoothed.

[Detailed Description of the Invention]

[0001]

[Technical Field of the Invention]

The present invention relates to an apparatus for suppressing noise components included in image signals carrying radiographic images, and to a storage medium for storing a program which instructs a computer to execute this noise suppression processing.

[0002]

[Description of the Related Art]

Currently, when radiographic images obtained by using a computed radiography device (hereinafter referred to as CR device) or the like are used in diagnosis, image processing such as frequency emphasis processing or gradation processing is performed on the obtained radiographic images before the

radiographic images are displayed on CRT monitors as soft copies or recorded in films as hard copies.

[0003]

Radiographic images suffer from the problem that they tend to include noticeable quantization noise in low density areas corresponding to low-intensity radiation exposure. Therefore, various methods have been proposed for suppressing noise components included in image signals carrying radiographic images.

[0004]

For example, Japanese Unexamined Patent Publications No. 6(1994)-96200 proposes a method of obtaining an image with noise components therein being suppressed, wherein an image is transformed (decomposed) into a set of detail images (represented by band-limited image signals corresponding to 1 to M resolution levels); a moving average of squared pixel values of each detail image in an N X N neighborhood centered around each pixel of interest (i.e., a sum of the squared pixel values divided by N^2) is calculated as a local variance; a local variance corresponding to the peak in the histogram of the local variance is obtained as a noise variance; the local variance corresponding to each pixel is compared with the noise variance; when the local variance is comparable to or smaller than the noise variance, a portion of the band-limited image signal corresponding to the pixel is reduced; and thereafter, the set of detail images processed as above are inversely composed to the space of the original image by inverse multiresolution transformation so that an image in which the noise is suppressed is obtained.

[0005]

Japanese Unexamined Patent Publications No. 6(1994)-96200 also discloses a technique for calculating noise variances for detail images at lower resolution levels based on a noise variance for a detail image at the highest resolution level (i.e., the finest-grained detail image).

[0006]

[Problems to be Solved by the Invention]

According to the method disclosed in Japanese Unexamined Patent Publications No. 6(1994)-274615 mentioned above, noise is suppressed by using a variance obtained from variances (moving average) only of a relevant resolution signal (detail image) and a histogram of the local variances, based on the assumption that the noise is uniformly distributed in the entire image. However, noise in actual radiographic images is not uniformly distributed. For example, noise levels are relatively high in areas in which images of objects exist, and noise levels are relatively low in areas in which images of objects do not exist. Therefore, when noise suppression is carried out using a threshold value calculated from the histogram, in the areas in which the noise levels are low, edge information is suppressed as well as the noise and thus edge degradation occurs, as a result of which sharpness of the radiographic images decreases, whereas in the areas in which the noise levels are high, it is not possible to sufficiently suppress the noise.

[0007]

In addition, histograms obtained from radiographic images including images of objects having complex structures and histograms obtained from radiographic images including images of objects having simple structures are different in shape. Therefore, it is difficult to desirably discriminate between edges and noise according to variations between objects.

[0008]

The present invention has been made in view of the points mentioned above. An object of the present invention is to provide an apparatus for effectively suppressing noise in or eliminating noise from a radiographic image regardless of an exposure dose with which the radiographic image has been produced, and reducing edge degradation which is likely to be

caused by noise suppression in a noisy radiographic image. Another object of the present invention is to provide a computer-readable storage medium storing a program which instructs a computer to conduct the noise suppression processing.

[0009]

[Means for Solving the Problems]

According to the present invention, there is provided a noise suppressing apparatus for suppressing noise components included in a radiographic image, comprising: smoothing means for processing an input image signal which represents an input radiographic image by using a smoothing filter such that the input image signal is smoothed; noise properties calculating means for calculating noise properties of the input image signal based on information indicating an exposure dose with which the radiographic image has been produced (hereinafter referred to as "exposure dose"), wherein the smoothing means adapts a characteristic of the smoothing filter which is used for performing the smoothing, based on the calculated noise properties.

[0010]

The information indicating an exposure dose with which the radiographic image has been produced may be information which directly indicates the exposure dose with which the radiographic image has been produced, for example, information from a photo-timer, or may be information which indirectly indicates the exposure dose with which the radiographic image has been produced, for example, information indicating a menu of radiography, the age of a patient, a condition of radiography (e.g., a condition for irradiation by a radiographic apparatus), a normalization condition (see, for example, Japanese Unexamined Patent Publication No. 2(1990)-108175), or a pixel value (density value) of the radiographic image.

[0011]

It is preferable that the noise suppressing apparatus of

the invention further comprises a band-limited-image-signal generation means for unit for generating a plurality of band-limited image signals respectively representing a plurality of band-limited images belonging to a plurality of different frequency bands, based on the input image signal, wherein the smoothing means processes the plurality of band-limited image signals by using the smoothing filter so as to smooth each of the plurality of band-limited image signals.

[0012]

It is preferable that the band-limited-image-signal generation means generates the plurality of band-limited image signals by performing multiresolution decomposition on the input image signal. In this case, a noise-suppressed band-limited image may be reconstructed by inverse multiresolution development. It is preferable that, for example, the Laplacian pyramid decomposition or wavelet transformation is used as the multiresolution transformation.

[0013]

As used herein, the term "based on information indicating an exposure dose with which the radiographic image has been produced" means "based on information indicating at least the exposure dose". It is of course that any other information may be used as well as the information indicating the exposure dose. The any other information may be, for example, local information (locally calculated information) in the vicinity of a pixel of interest. That is, in the noise suppressing apparatus of the invention, it is preferable that the noise properties calculating means obtains the noise properties also based on information locally calculated from pixel values in the vicinity of a pixel of interest in at least one of the plurality of band-limited image signals.

[0014]

The "locally calculated information" may be any information from which a degree of likelihood that the pixel constitutes an edge can be verified. For example, a local

average (or moving average deviation) of density values and vector information, or a local square-sum may be used.

[0015]

In this case, it is preferable that the noise properties calculation means obtains a pixel vector at the pixel of interest in a band-limited image represented by at least one of the band-limited image signals, and detects an orientation of an edge as the noise properties using the pixel vector as the locally calculated information, and the smoothing means adapts a characteristic of the smoothing filter so that the smoothing is performed along the orientation of the detected edge.

[0016]

According to the present invention, there is provided a computer-readable storage medium storing a program which instructs a computer to execute on an input image signal representing an input radiographic image a process for suppressing noise components included in the radiographic image, wherein the program instructs the computer to carry out the steps of: calculating noise properties of the input image signal based on information indicating an exposure dose with which the radiographic image has been produced; adapting based on the calculated noise properties a characteristic of a smoothing filter used for processing the plurality of band-limited image signals such that each of the band-limited image signals is smoothed; and processing the input image signal by using the adapted smoothing filter such that the input image signal is smoothed.

[0017]

According to the present invention, there is provided computer-readable storage medium storing a program which instructs a computer to execute on an input image signal representing an input radiographic image a process for suppressing noise components included in the radiographic image, wherein the program instructs the computer to carry out

the steps of: generating a plurality of band-limited image signals respectively representing a plurality of band-limited images belonging to a plurality of different frequency bands, based on the input image signal; calculating noise properties of the input image signal based on information indicating an exposure dose with which the radiographic image has been produced; adapting based on the calculated noise properties a characteristic of a smoothing filter used for subjecting the input image signal to smoothing; and processing the input image signal by using the adapted smoothing filter such that the input image signal is smoothed.

[0018]

[Advantageous effect of the Invention]

According to the noise suppressing apparatus and the storage medium of the present invention, noise properties of the input image signal is calculated based on information (exposure dose information) indicating an exposure dose with which the radiographic image has been produced, and a characteristic of the smoothing filter is adapted based on the so calculated noise properties. Therefore, an image in which noise is effectively suppressed or eliminated can be obtained regardless of the exposure dose.

[0019]

Further, when band-limited image signals are generated and processed by using a smoothing filter so as to smooth each of the band-limited images, only one image signal is need to be reconstructed based on each of the noise-suppressed band-limited image signal. Also in this case, an image in which noise is effectively suppressed or eliminated can be obtained regardless of the exposure dose.

[0020]

Further, when noise properties is obtained based on information locally calculated from pixel values in the vicinity of a pixel of interest in at least one of the plurality of band-limited image signals, the noise properties taking into

account the degree of the edge and the exposure dose information, and separation between the edge and noise based on the noise variation for each area becomes easy. As a result, an effectively-noise-suppressed image with less edge degradation can be obtained.

[0021]

In this case, when a pixel vector at the pixel of interest in a band-limited image is obtained, an orientation of an edge as the noise properties is detected using the pixel vector as the locally calculated information, and a characteristic of the smoothing filter is adapted so that the smoothing is performed along the orientation of the detected edge, edge degradation which is likely to be caused by noise suppression in a noisy radiographic image can be reduced and variance in image quality due to variations in the exposure dose can be suppressed. In other words, since the smoothing can be performed along the orientation of an edge, noise on the edge can be properly suppressed and noise on the edge can be more effectively suppressed without reducing edge contrast. Thus, the noise can be effectively suppressed while preserving the edge.

[0022]

Further, when a plurality of filters respectively smoothing the radiographic image in the predetermined direction to a plurality of different degrees is provided in advance, and characteristics of the smoothing filter are changed by selecting one of the plurality of filters based on the calculated noise properties, processing for noise suppression can be performed in a short time, and therefore, the overall throughput of the processing can be increased.

[0023]

Further, when the band-limited image signals are generated from the input image signal by multiresolution decomposition, the generation of the plurality of band-limited image signals and the reconstruction of the noise-suppressed image can be performed in a short time. Therefore, the overall

throughput of the noise suppression processing can be further increased.

[0024]

[Embodiments of the Invention]

Hereinafter, embodiments of the present invention will be described in detail with reference to the drawings.

[0025]

Fig. 1 is a diagram illustrating an outline of a construction of a noise suppressing apparatus according to the present invention.

[0026]

As shown in FIG. 1, a noise suppressing apparatus 100 comprises: a band-limited-image-signal generation unit 1, the band-limited-image-signal generation unit 1 generating a plurality of band-limited image signals that represent a plurality of band-limited images respectively belong to a plurality of different frequency bands, based on an input image signal Sin that represents a radiographic image obtained by an image reading device or the like and having a certain resolution; an index-value obtaining unit 2, the index-value obtaining unit 2 obtaining at least one index value which indicates a degree of noise suppression based on information indicating an exposure dose with which the radiographic image has been produced; a noise-suppression processing unit 3, the noise-suppression processing unit 3 performing noise suppression processing on each of the plurality of band-limited image signals according to the degrees of noise suppression; and an image reconstruction unit 4, the image reconstruction unit 4 composing (reconstructing) a processed image signal Sproc which represents a noise-suppressed radiographic image, from the plurality of band-limited image signals on which the noise suppression processing is performed by the noise-suppression processing unit 3.

[0027]

The index-value obtaining unit 2 is configured to serve

as noise properties calculating means of the invention. That is, the index-value obtaining unit 2 obtains a pixel vector at each pixel of interest in the band-limited image, and detects an orientation of an edge as noise properties by using the pixel vector.

[0028]

The noise-suppression processing unit 3 is configured to serve as a smoothing means of the present invention. That is, the noise-suppression processing unit 3 adapts characteristics of smoothing filters so that smoothing processing can be performed along the orientation of the edge calculated (or detected) by the index-value obtaining unit 2 and the smoothing processing is performed on each of the plurality of band-limited image signals Bk by use of the smoothing filter after the adaptation of characteristics is changed.

[0029]

Further, according to the particular embodiment, for example, in a radiographic-image-information recording-and-reproducing system as disclosed in Japanese Unexamined Patent Publication Nos. 55(1980)-12492 and 56(1981)-11395, a radiographic image of a human body that is recorded in a stimulable phosphor sheet and read as a digital image signal by laser beam scanning is targeted for the noise suppression processing. The radiographic image is read by performing the laser beam scanning in two dimensions by moving a laser beam on the stimulable phosphor sheet in a main scanning (lateral) direction while moving the stimulable phosphor sheet in a feeding (longitudinal) direction.

[0030]

Operations of a noise suppressing apparatus 100 having the structure described above are explained below.

[0031]

First, an outline of the processing is described with reference to a flowchart shown in FIG. 2.

[0032]

In order to generate the plurality of band-limited image signals, it is preferable to use the multiresolution transformation such as the Laplacian pyramid decomposition that is proposed in Japanese Unexamined Patent Publication Nos. 5(1993)-244508 and 6(1994)-096200 and Japanese Patent Application Nos. 11 (1999)-363766 and 2000-022828 or the wavelet transformation that is proposed in Japanese Unexamined Patent Publication Nos. 6(1994)-274615 and Japanese Patent Application 11-363766 which are also assigned to the assignee of the present invention. Alternatively, the plurality of band-limited image signals may be generated by using other known methods. For example, the plurality of band-limited image signals may be generated by using the unsharp mask signals as disclosed Japanese Unexamined Patent Publication No. 10(1998)-75364. Descriptions on embodiments below are based on the assumption that the Laplacian pyramid decomposition is used.

[0033]

Band-limited image signals are generated from an entered original image by the use of the Laplacian pyramid decomposition that is one of the multiresolution transformations (step S21). Then, vector components at each pixel position of each of a plurality of band-limited images in multiresolution spaces are calculated, where the band-limited images are respectively represented by the band-limited image signals (step S22). When the vector components are obtained in double-angle representation (which is explained later), four-orientation vector components, i.e., vector components corresponding to four orientations at intervals of 45 degrees, are obtained in each pixel position. Based on the four-orientation vector components, it is possible to discriminate between noise components and edge components in each pixel position.

[0034]

When a singular point (local noise) exists, e.g., when a vector which is extremely greater than surrounding vectors

exists, the local noise at each pixel position is likely to be incorrectly recognized as an edge signal. Therefore, the vicinity average (vector average), i.e., an average of values of each vector component in the neighborhood, is obtained by using a one-dimensional filter (step S23). The vector average is based on the assumption that edge signals are continuous. In this embodiment, the neighborhood average is obtained by using an isotropic two-dimensional space filter. Further, the vector average is modified in by using a vector component obtained at a resolution level lower than the resolution level at which the neighborhood average is obtained. At this time, the modification of the vector average is made based on information on the exposure dose with which the original radiographic image has been produced (step S24).

[0035]

Next, a degree C of edge confidence and an index E of pixel energy are calculated based on each vector averaged and modified as above in accordance with a method explained later (step S25). Noise suppression processing using adaptive filtering is performed based on the degree C of edge confidence and the index E of pixel energy (step S26). Finally, Laplacian pyramid reconstruction that is one of the inverse multiresolution transformations is made so that a processed image in which noise is suppressed is obtained (step S27).

[0036]

The adaptive filtering in step S26 is performed by an anisotropic filter (orientation-dependent filter) and an isotropic filter (orientation-independent filter). It is possible to calculate tens of different sets of anisotropic filter coefficients for the anisotropic filter in advance, and select one of the tens of different sets according to a vector orientation D and an amount of noise. On the other hand, the isotropic filter can be realized by a simple non-linear transformation.

[0037]

In addition, when an isotropic filter is produced, one coefficient at the center of a mask is placed at one, and the other filter coefficients are calculated in accordance with the following equation (1). Then, the filter coefficients are normalized so that the sum of the filter coefficients is equal to one.

[Formula 1]

$$\left.\begin{aligned} &F_{i,j} = \exp\{-\pi \times f(x) \times (i^2 + j^2)/(N \times N \times 2)\} \\ &-N \le i \le N \; and \; -N \le j \le N \end{aligned}\right\} \quad (1)$$

[0038]

On the other hand, the filter coefficients of the anisotropic filter are calculated in accordance with the following equations (2). Then, the filter coefficients are normalized so that the sum of the filter coefficients is equal to one.

[Formula 2]

$$\left.\begin{aligned} &X_{i,j} = [\cos\{\cos^{-1}(i / \mathrm{sqrt}(i^2 + j^2))\, deg\}]^2 \\ &-N \le i \le N \\ &-N \le j \le N \\ &Y_{i,j} = X^{u(x)} \times F_{i,j} \end{aligned}\right\} \quad (2)$$

where deg denotes an angle from a vector orientation.

[0039]

Next, details of the operations performed in the sequence are explained below.

[0040]

Fig. 3 is a block diagram illustrating an outline of a construction of the band-limited-image-signal generation unit 1 and schematically illustrating the operations of generating five band-limited image signals corresponding to five resolution levels.

[0041]

For example, as disclosed in Japanese Unexamined Patent Publication No. 5(1993)-244508, the filtering processing unit

10 performs filtering processing on the input image signal Sin in each of the main scanning and feeding directions so as to produce a signal L1 (hereinafter referred to as a lower resolution signal), which belongs to a lower resolution level than the resolution level of the input image signal Sin. Then, the filtering processing unit 10 performs filtering processing on the lower resolution signal L1 in each of the main scanning and feeding directions so as to produce a next lower resolution signal L2. Thereafter, further lower resolution signals Lk (k=1 to n) are successively obtained by repeating the filtering processing in a similar manner. Next, the interpolation processing unit 11 performs interpolation processing on each of the lower resolution signals Lk so that the number of pixels in each of the main scanning and feeding directions is doubled, i.e., the number of pixels in the lower resolution signal Lk is increased by a factor of four. Thus, a plurality of unsharp image signals Sus1 to Susn (hereinafter collectively referred to as "Susk (k=1 to n)") each having a different degree of sharpness are obtained. Thereafter, the subtractor 12 obtains a difference between each of the lower resolution signal Lk-1 and one of the plurality of unsharp image signals Susk having the same number of pixels as the lower resolution signal Lk-1 so as to generate one of the band-limited image signals Bk.

[0042]

Next, details of the operations performed for obtaining an index value (a degree of noise suppression) and suppressing noise by using the band-limited image signals Bk obtained as described above are explained below.

[0043]

Fig. 5 is a block diagram illustrating details of a general configuration of the noise suppressing apparatus 100. As shown, the index-value obtaining unit 2 comprises for each of the plurality of band-limited image signals Bk: a pixel-vector generation unit 22 for generating a pixel vector at each pixel of a band-limited image represented by the

corresponding band-limited image signal Bk; and an index-value calculation unit 24 for obtaining at least one of a degree of edge confidence, an index of pixel energy, and an edge orientation (which are examples of the index values of the present invention), for each pixel of the band-limited image represented by the corresponding band-limited image signal Bk, based on the length and/or orientation of the pixel vector generated by the pixel-vector generation unit 22 corresponding to the same resolution level.

[0044]

The noise-suppression processing unit 3 comprises a suppression processing unit 32 for each of the plurality of band-limited image signals Bk. The suppression processing unit 32 serves to suppress noise components included in the band-limited image signal Bk, based on the index value output from the index-value calculation unit 24.

[0045]

The noise suppression processing in the present embodiment is based on a technique that "a line signal is smoothed along the orientation of the line, and isolated noise is two-dimensionally smoothed". The most characteristic feature of the noise suppression processing is that a smooth edge is obtained by the smoothing operation of the line (edge) signal, and information necessary for the smoothing operation is represented in only a vector or tensor form. In the examples disclosed in this specification, the double-angle representation (hereinafter referred to as "D-A representation") representation is used as a vector representation form.

[0046]

The D-A representation of a vector is a technique for representing a line signal, and advantageous in that the degree of confidence of a line signal (index of lineness) can be obtained by only calculating the vicinity average of information in the D-A representation. This feature is

explained below with reference to FIG. 6.

[0047]

When density vectors of an image signal as shown in FIG. 6(A) are calculated, the density vectors are represented as illustrated in FIG. 6(B) in the full-angle representation (hereinafter referred to as "F-A representation"), which is an example of normal vector representations. Thus, the orientations of the vectors on the respective sides of the boundary are opposite, where the low-density area in the image corresponds to the boundary. On the other hand, when represented in the D-A representation, the orientations of the vectors on both sides of the boundary are identical as illustrated in FIG. 6(C) since the angle values of the vectors are doubled in the D-A representation.

[0048]

In addition, when the degrees C of edge confidence of the above vectors are obtained by the neighborhood average as indicated with the bold arrows in FIGS. 6(B) and 6(C), the degree C of edge confidence shown in FIG. 6(C) is considerably smaller than the degree C of edge confidence shown in FIG. 6(B). Although not specifically shown, it will be easily understood that the confidence of noise also becomes small (since vectors surrounding a vector of interest have random orientations). Therefore, it is difficult to discriminate between noise and line information in the F-A representation.

[0049]

On the other hand, in the D-A representation, vectors indicating line (edge) orientations can be defined as illustrated in FIG. 7. In the drawing, q0 to q3 each indicate the magnitude of a directional component at a pixel of interest. When the magnitudes of two orthogonal directional components at a pixel are identical (i.e., when the pixel is located at a point of intersection of two lines), the output in the D-A representation becomes small. When two orthogonal directional components have different magnitudes, the orientation

corresponding to the greater magnitude becomes a main orientation at the pixel.

[0050]

Therefore, when the four directional components q0 to q3 of a vector at each pixel are obtained, the vector can be represented in the D-A representation.

[0051]

Details of operations for obtaining the directional components q0 to q3 are specifically described below.

[0052]

The band-limited image signals used for calculation are Laplacian signals generated by the Laplacian pyramid decomposition. The four directional components are calculated by convolution of the Laplacian signals with the four two-dimensional filters as illustrated in FIG. 8. Table 4-1 shows examples of filter coefficients of a 5 X 5 q0 filter. Since each of the Laplacian signals can become positive or negative, and each of the filter coefficients can be positive or negative, absolute values of the convolution products are used for calculation of the directional components.

[TABLE 1]

Filter Coefficients for q0 Filter (5-by-5)

0.0012	0.0211	0.0577	0.0211	0.0012
0.0053	0.1389	0.6093	0.1389	0.0053
0.0000	0.0000	0.0000	-0.0000	-0.0000
-0.0053	-0.1389	-0.6093	-0.1389	-0.0053
-0.0012	-0.0211	-0.0577	-0.0211	-0.0012

[0053]

Each of the above-mentioned four filters is a kind of differential filter. Therefore, in the case where a second-derivative signal such as a Laplacian signal is convoluted, the output of the filter does not become great in regions in which the gradient of the second-derivative signal is not great, even when the second-derivative signal per se is

great in the regions. This feature is explained below with reference to FIG. 9.

[0054]

FIG. 9 is a diagram illustrating a relationship between a Laplacian signal and an output of a first-derivative filter (the absolute value of the first derivative of the Laplacian signal). As shown in FIG. 9, the point a of the Laplacian signal corresponds to a portion of an edge region, and the Laplacian signal is maximized at the point A. However, since the gradient of the Laplacian signal at the point A is zero, the output of the first derivative corresponding to the point a becomes zero.

[0055]

In addition, at a point C which is a boundary point between the edge region and a non-edge region, the output of the first derivative corresponding to the point C becomes grater than the output of the first derivative corresponding to the point A. This tendency is strengthened with increase of the mask size. However, when the mask size is increased, trackability of very small edges decreases, and therefore sharpness of images of the very small edges decreases.

[0056]

Thus, a small mask size is preferable when the image signal does not include artificially produced noise as in a pattern image. In an actual input image as it is, however, noise exists in the entire image in varying degrees. Therefore, first derivatives generated by using filters having a small mask size are strongly affected by noise.

[0057]

The above circumstances can be a cause of image quality degradation when an adaptive filter which operates based on pixel energy (i.e., an average of the directional components q0 to q3) is used.

[0058]

Nevertheless, when the amount of noise can be estimated, it is possible to optimize the filter by changing the mask size

according to the amount of noise, or setting filter coefficients so as to substantially achieve the effect of the setting of the mask size. For example, when a radiographic image is to be processed as in the present embodiment, an X-ray dose and an amount of noise can be estimated from the S value (indicating the reading sensitivity) and the L value (indicating the latitude), and an optimum set of filter coefficients can be calculated. Regarding the S value and the L value, see, for example, Japanese Unexamined Patent Publication No. 2(1990)-108175.

[0059]

Specifically, the vicinity average of each of the vector components (q0 to q3) at each pixel is obtained. For the neighborhood average, an isotropic two-dimensional filters as illustrated in FIG. 10 is used.

[0060]

When the mask size of the two-dimensional filter is varied, the smoothing level of the vector component of course varies. The smoothing level is reflected on the edge confidence and the pixel energy, and the influence of the smoothing level on the final image is relatively great. When the mask size is increased, noise and relatively large edges can be discriminated with high accuracy, and small edges are likely to be regarded as noise. Therefore, smoothing with a large mask size is effective when the input image does not include fine structures, for example, as in a radiographic image of a chest of a child. On the other hand, since bone images such as an image of a foot includes complex fine structures such as trabecula here and there, fine signals indicating the complex fine structures cannot be recognized when the smoothing level is raised. Therefore, a small mask size is used.

[0061]

As proposed in Japanese Patent Application No. 2000-022828 by the present applicant, the above neighborhood average can be modified by using vector components in a lower

resolution image which has a lower resolution than the targeted image. Investigations by the present inventors have revealed the following characteristics when the vector calculated from a currently targeted band-limited image signal (band-limited image of interest) for vector averaging and another vector (having a lower resolution) are used.

1) An image including therein a smaller amount of noise (the image which undergoes a large amount of X-ray exposure) has a higher signal-to-noise ratio. Therefore, vector components obtained by the use of the vector average of the currently targeted band-limited image signals can more faithfully follow fine signals representing fine structures in the input image than vector components modified with the lower resolution signals, which prevents the edge degradation.

2) An image including therein a larger amount of noise has a lower signal-to-noise ration. Therefore, when an average of a vector calculated from a band-limited image signal in a frequency band (with a lower SNR) of interest and another vector calculated based on image information in a lower-frequency band (with an enhanced SNR) is used, the average of the vectors can more faithfully follow relatively large signals which are not buried in the noise in the input image, and noise can be effectively suppressed.

[0062]

Thus, in order to improve image quality, it is important to control the degree of noise suppression according to the amount of noise included in the input image.

[0063]

Then, a noise amount estimating process which becomes necessary for controlling the degree of noise suppression is described below.

[0064]

For a radiographic image such as an X-ray photogram, noises are mainly caused by reduction of transmission radiation dose. Therefore, if the transmission radiation dose is known,

an approximate noise amount could be estimated.

[0065]

The average C of vectors having respectively different frequency bands can be calculated in accordance with the equation (3) given below, where A is a average of vectors (vector average) calculated from a band-limited image signal in a frequency band of interest, B is another average of vectors calculated based on image information in a lower frequency band.

[Formula 3]

$$C = f(x) \times A + (1 - f(x)) \times B \qquad (3)$$

[0066]

In the equation, x represents an X-ray dose, and f(x) is a function of the X-ray dose x, and represents a weight of the neighborhood-averaged vector in the weighted average.

[0067]

For estimating the amount of noise, various information which may indicate X-ray dose can be used, for example, 1) the radiographed region or the menu of radiography; 2) the S or L value indicating a normalization condition (EDR condition); 3) the signal values (density values) of the image; and 4) the age of a patient or the condition of radiography.

[0068]

When the radiographed region or the menu of radiography is used, for example, a low-dose menu, a child menu, and the like may be provided, and the weights of the aforementioned neighborhood-averaged vectors in the weighted average can be changed according to a selected one of the menu. When the S or L value indicating a normalization condition is used, it is possible to use the function f(x) which increases with decrease in the S value (corresponding to increase in the X-ray dose). For example, it is preferable to use, for example, the functions f(x) given by the following set of equations (4). It is of course that other such functions may be used.

[Formula 4]

$$\left.\begin{array}{l} f(x) = 1.0 \text{ when } S < 100 \\ f(x) = (2000 - S) / 1900 \text{ when } 100 \leq S \leq 2000 \\ f(x) = 0.0 \text{ when } S > 2000 \end{array}\right\} \quad (4)$$

[0069]

When the signal value of the image is used, since the density value corresponds to the X-ray exposure on a stimulable phosphor sheet, the relative amount of the X-ray exposure dose can be indicated by using the signal value. Therefore, it is preferable to use the functions, for example, as given by the following set of equations (5).

[Formula 5]

$$\left.\begin{array}{l} x = S \times 10^{(-L \times QL / 1024)} \\ f(x) = 1.0 \text{ when } x < 100 \\ f(x) = (2000 - x) / 1900 \text{ when } 100 \leq x \leq 2000 \\ f(x) = 0.0 \text{ when } x > 2000 \end{array}\right\}$$

(5) wherein QL denotes a signal value, and x denotes a relative amount of the X-ray dose.

[0070]

Alternatively, the set of equations (5) may be modified so that the function f(x) satisfies $0.5 \leq f(x) \leq 1.0$, instead of $0 \leq f(x) \leq 1.0$.

[0071]

That is, in the case where the signal values are used, the signal value of each pixel is referred to when the vector average is calculated, the relative amount of the X-ray exposure dose is estimated in accordance with the set of equations (5), and a weighted average of vectors is calculated by using the weights of the neighborhood-averaged vectors determined in accordance with the function defined in advance.

[0072]

Next, the orientations and lengths of a primary vector and a secondary vector are calculated by using the four vector components (q0 to q3). Since each vector component is obtained in the D-A representation as illustrated in FIG. 7, the length V1 of the primary vector is calculated in accordance with

equations (6), and the unit-vector components ex1 and ey1 of the primary vector are calculated in accordance with the equations (7).

[Formula 6]

$$\left.\begin{aligned} Z1 &= q0 - q2 \\ Z2 &= q1 - q3 \\ V1 &= \left(z1^2 + z2^2\right)^{1/2} \end{aligned}\right\} \qquad (6)$$

[Formula 7]

$$\left.\begin{aligned} ex1 &= z1 / V1 \\ ey1 &= z2 / V1 \end{aligned}\right\} \qquad (7)$$

[0073]

Since the orientation of the secondary vector is opposite to that of the primary vector in the D-A representation, the unit-vector components (ex2, ey2) of the secondary vector oriented orthogonal to the primary vector are calculated in accordance with equations (8).

[Formula 8]

$$\left.\begin{aligned} ex2 &= -ex1 \\ ey2 &= -ey1 \end{aligned}\right\} \qquad (8)$$

[0074]

In addition, the amount of pixel energy Ve at each pixel is defined as an average of the components in accordance with the equations (9). Further, the length V2 of the secondary vector can be calculated from the ratio of the pixel energy Ve and the length V1 of the primary vector in accordance with the equations (10)

[Formula 9]

$$Ve=(q0+q1+q2+q3)/4 \qquad (9)$$

[Formula 10]

$$V2=(1-V1/Ve)\times V1 \qquad (10)$$

[0075]

Next, by using the two pieces of information V1 and V2 on the mutually orthogonal primary and secondary vectors which have been averaged and modified based on the noise amount, the

degree C of edge confidence, the index E of pixel energy, and a smoothing direction D of the anisotropic filter are calculated in accordance with the following equations. That is, the index E of pixel energy is calculated in accordance with the set of equations (11) by using a predetermined threshold value Th, and the degree C of edge confidence is calculated from the lengths V1 and V2 of the primary and secondary vectors in accordance with the equation (12).

[Formula 11]

$$\left.\begin{array}{l} E = (Ve / Th)^2 / 2 \quad \text{when } Ve < Th \\ E = \{1 - (2 - Ve / Th)^2\} / 2 \quad \text{when } Th \le Ve < 2 \times Th \\ E = 1.0 \quad \text{when } 2 \times Th \le Ve \end{array}\right\}$$

(11)

[Formula 12]

$$C=(V1-V2)/V1 \qquad (12)$$

[0076]

Note that, $0.0 \le E \le 1.0$, $0.0 \le C \le 1.0$, $0.0 \le D \le 31$ (D is an integer).

[0077]

Further, as shown in the set of equations (13), an angle θ is calculated from the unit-vector components of the secondary vector, and a quantized smoothing direction D of the anisotropic filter is calculated from the calculated angle. In this particular embodiment, 32 discrete angle values are employed.

[Formula 13]

$$\left.\begin{array}{l} \theta = \cos^{-1}(ex2) \quad when\ ey2 > 0 \\ \theta = \cos^{-1}(-ex2) \quad when\ ey2 \le 0 \\ D = f(\theta) \end{array}\right\} \qquad (13)$$

wherin f(θ) is the function that converts the continuous angle into discrete angle values.

[0078]

The degree C of edge confidence indicates a degree of

likelihood that the pixel constitutes an edge, and becomes larger as the relevant pixel is positioned closer to a line and the larger index E of pixel energy indicates a degree of likelihood that the pixel constitutes a signal. A recognition model (an example of an adaptive filter) of a line, a point of intersection, an end point, and noise based on the degree C of edge confidence and the index E of pixel energy are indicated in FIG. 11.

[0079]

Next, smoothing processing using an anisotropic filter and adaptive filtering are performed on each band-limited image signal (Laplacian signal) at each pixel based on the degree C of edge confidence, the index E of pixel energy, and the smoothing direction D, which are obtained as above from the information on the two mutually orthogonal vectors.

[0080]

In the present embodiment, smoothing processing is performed by using an anisotropic filter which is oriented along the line recognized based on the degree C of edge confidence since the larger the degree C of edge confidence, the greater the likelihood that a line is constituted. That is, the two-dimensional anisotropic spatial filter (orientation-dependent filter) smoothes the Laplacian signal along the orientation of the primary vector. The coefficients of the anisotropic filter are set to have a shape, for example, as illustrated in FIG. 12. Since noise in a noisy image is also superimposed on edge signals, the anisotropic filter as above is used for suppressing noise on an edge without reducing edge contrast.

[0081]

For the respective anisotropic filters shown in FIG. 12, greater coefficients are indicated with higher density. A filter (a) shown on the left side as viewed in the drawing is for performing smoothing processing in the vertical direction. Although the smoothing capability is increased with increase

in the center angle θ, smoothing of the edge signal is enhanced with the increase in the center angle θ. In addition, while the filter shown at the higher position in the drawing has a larger mask size and therefore has a higher smoothing capability, edge degradation is enhanced. Conversely, while the filter shown at the lower position in the drawing has a smaller mask size and therefore has a lower smoothing capability, the edge can be preserved.

[0082]

When two-dimensional anisotropic filters as described above are used, it is possible to perform smoothing processing along the vector orientations. That is, noise on edges can be suppressed or eliminated while preserving the edges. Investigations by the present inventors have revealed that the optimum extent of an effective area (mask size) of the anisotropic filter and the optimum degree of directionality (center angle) of the anisotropic filter vary with the amount of noise included in the image, as described below. That is, in order to suppress or eliminate noise on an edge while preserving the edge, it is important to adaptively change the characteristics of the anisotropic filter according to the amount of noise included in the image.

1) When the amount of noise included in the image is small (when an X-ray exposure dose with which the input image has been produced is high), it is preferable to use a filter which has a small effective area and a small center angle so that fine signals can be followed more faithfully.

2) When the amount of noise included in the image is great, if such an image is processed under the same condition as that mentioned in 1), artifacts of the noise oriented along the orientation of the vector are produced. Therefore, higher filtering characteristics involving smoothing capability are required.

[0083]

Then, a noise amount estimating process which becomes

necessary for changing the anisotropic filter is described.

[0084]

For a radiographic image such as an X-ray photogram as described above, noises are mainly caused by reduction of transmission radiation dose. Therefore, if the transmission radiation dose f(x) is known, an approximate noise amount could be estimated.

[0085]

For estimating transmission radiation dose f(x), various information which may indicate X-ray dose can be used, for example, 1) the radiographed region or the menu of radiography; 2) the S or L value indicating a normalization condition (EDR condition); 3) the signal values (density values) of the image; 4) the degree C of edge confidence; and 5) the age of a patient or the condition of radiography.

[0086]

When the radiographed region or the menu of radiography is used, for example, a low-dose menu, a child menu, and the like may be provided, and it is preferable to set the function f(x) in advance for each menu.

[0087]

When the S or L value indicating a normalization condition is used, it is possible to use the function f(x) which increases with decrease in the S value (corresponding to increase in the X-ray dose). For example, it is preferable to use, for example, the functions f(x) given by the following set of equations (14). It is of course that other such functions may be used.

[Formula 14]

$$\left.\begin{array}{ll} f(x)=10-3\times\log S & \textit{when}\ 10-3\times\log S\geq 1 \\ f(x)=1 & \textit{when}\ 10-3\times\log S<1 \end{array}\right\} \qquad (14)$$

[0088]

When the signal value of the image is used, since the density value corresponds to the X-ray exposure on a stimulable phosphor sheet, the relative amount of the X-ray exposure dose

can be indicated by using the signal value. Therefore, it is preferable to use the functions as given by the following set of equations (15) for defining the transmission radiation dose f(x).

[Formula 15]

$$\left.\begin{array}{ll} x = S \times 10^{(-L \times QL/1024)} & \\ f(x) = 10 - 3 \times \log x & when\ 10 - 3 \times \log x \geq 1 \\ f(x) = 1 & when\ 10 - 3 \times \log x < 1 \end{array}\right\} \qquad (15)$$

wherein QL denotes a signal value, and x denotes a relative amount of the X-ray dose.

[0089]

When filtering is performed using these equations, the value of the function f(x) is obtained with reference to the image position at the relevant pixel position, and the characteristics of the anisotropic filter is changed based on the obtained value of the function f(x). Accordingly, when the signal value is small (i.e., the X-ray exposure dose is low), the center angle and the mask size of the anisotropic filter are increased, and therefore the smoothing capability is increased. Conversely, when the signal value is great (i.e., the X-ray exposure dose is high), the center angle and the mask size of the anisotropic filter are decreased, and therefore the edge degradation can be prevented.

[0090]

As described before, the degree of edge confidence is the index obtained after the aforementioned average of the vectors is obtained, and each pixel is less likely to constitute noise when the value is greater. Conversely, the more the noise in an image, the lower the degrees of edge confidence. Therefore, the degree of edge confidence can also be used as an index for estimating the amount of noise. That is, when the degree of edge confidence is great, the center angle and the mask size of the smoothing filter is decreased, while when the degree of edge confidence is small, the center angle and the mask size of the

smoothing filter is increased, whereby it is possible to concurrently realize the noise suppression and the edge sharpening.

[0091]

Further, although anisotropic filters preserve edges, the anisotropic filters smooth points at which edges intersect. However, intersection points appearing on actual images are not ideal intersection points (wherein sharp lines intersect at the right angle), and therefore are not so smoothed.

[0092]

Next, an anisotropic filter as an orientation-dependent filter is selected depending on the smoothing direction $D(\theta)$ and the amount of noise both of which are calculated based on vectors. Then, the band-limited image signal is convoluted with the selected filter so as to produce a convolution product as an anisotropic-filtered signal. The convolution product (anisotropic-filtered signal) is designated by A in equation 16 and equation 17.

[0093]

As adaptive filtering for suppressing the noise components, the noise-suppression processing unit 3 calculates, as described above, a convolution product A of the anisotropic-filtered signal and the Laplacian signal and controls the weight of the convolution product A and the Laplacian signal based on the index E of pixel energy and the degree C of edge confidence so as to obtain a processed band-limited image signal fBk (k=1 to n) (represented by Proc in the equations) for each pixel of the band-limited image represented by the band-limited image signal (Laplacian signal). The processed band-limited image signal Proc is a signal in which noise components are suppressed. Definitions applied when calculating the band-limited image signal Proc are Definitions 1 and 2 described below.

[0094]

Definition 1: When FIG. 11(b) is defined as a linear:

This is the case where FIG. 11(b) is defined as a linear, and the processed band-limited image signal Proc in which noise components are suppressed is calculated in accordance with the following equation (16).

[Formula 16]

$$\text{Proc}=C\times A+E\times(1-C)\times Org\,, \qquad (16)$$

[0095]

When the degree C of edge confidence is high, an anisotropic filter output is selected because of edges (first term), while when the degree C of edge confidence is low, the original Laplacian signal is attenuated by the index E of pixel energy. Thus, noise and intersection points are separated from edges (second term).

[0096]

Definition 2: When FIG. 11(b) is defined as a noise:

This is the case where FIG. 11(b) is defined as a mpose, and the processed band-limited image signal Proc in which noise components are suppressed is calculated in accordance with the following equation (17).

[Formula 17]

$$\text{Proc}=E\times C\times A+E\times(1-C)\times Org\,. \qquad (17)$$

[0097]

When the degree C of edge confidence is high, the anisotropic-filtered signal is attenuated by the index E of pixel energy, so that the noise and edges are separated from each other (first term). When the degree C of edge confidence is low, the Laplacian signal is attenuated by the index E of pixel energy, so that noise and intersection points are separated from each other (second term).

[0098]

However, in the case of Definition 1, noise can be eliminated from the processed band-limited image signal fBk (Proc) only when the signal-to-noise ratio (SNR) is high. However, when the signal-to-noise ratio is low, noise elimination effect is not substantially produced. From these

results, it is considered the degree of edge confidence at a pixel can become high in the low-SNR situation even when the pixel represents noise.

[0099]

In the case of Definition 2, an artifact resulting from discontinuity (local discontinuity caused by noise or an artificially produced edge which suddenly appears) is more likely to be produced as the signal-to-noise ratio decreases. In order to eliminate the artifact, the threshold value used in the calculation of pixel energy is required to be raised. However, when the threshold value is raised, edges become unclear. This is because the pixel energy of edge competes with the pixel energy of noise in the low-SNR situation. This problem is inevitable as long as the pixel energy is used.

[0100]

Further, the degree C of edge confidence of a pixel has a small value when the pixel constitutes a nonlinear signal (representing an intersection point, an end point, or the like). Therefore, in order to discriminate between the nonlinear signal and noise, it is possible to determine whether each pixel corresponds to a true signal or noise by comparing the index E of pixel energy with a predetermined value. In practice, it is possible to continuously determine the unlikelihood of noise by using an arbitrary nonlinear function. For example, the unlikelihood N of noise can be determined by using a nonlinear function of a threshold value TH and the index E of pixel energy, as indicated in the equation (18).

[Formula 18]

$$N = \frac{\exp(TH / E) - 1}{\exp(TH / E) + 1} \times 2 \times \frac{E}{TH} \qquad (18)$$

[0101]

In this case, it is preferable to replace the aforementioned equation (16) with the equation (19), and the aforementioned equation (17) with the equation (20) based on the degree C of edge confidence and the unlikelihood N of noise.

[Formula 19]

$$\text{Proc}=C\times A+N\times(1-C)\times Org \qquad (19)$$

[Formula 20]

$$\text{Proc}=E\times C\times A+N\times(1-C)\times Org \qquad (20)$$

[0102]

In the equation (18), the threshold value TH is a value determined based on an amount corresponding to an exposure dose. When the amount corresponding to the exposure dose is a pixel value, the threshold value TH varies from pixel position to pixel position. In addition, the function defined by the equation (18) asymptotically approaches 1.0 with the increase of energy. For example, according to equation (19), when the index E of pixel energy is sufficiently greater than the threshold value TH, the unlikelihood N of noise becomes nearly equal to one and a processed image signal Sproc remains, while when the index E of pixel energy is sufficiently smaller than the threshold value TH, the unlikelihood N of noise becomes nearly equal to zero and therefore the processed image signal Sproc becomes equal to zero.

[0103]

Thus, after the noise suppression processing is performed on the band-limited image signals, i.e., the processed band-limited image signals Proc are obtained, the image reconstruction unit 4 makes a Laplacian reconstruction as an inverse multiresolution transformation so as to produce a processed image signal Sproc which represents an image in which noise components are suppressed.

[0104]

As illustrated in FIG. 5, the image reconstruction unit 4 comprises an interpolation processing unit 43 and an adder 44 corresponding to each resolution level. The interpolation processing unit 43 performs interpolation processing on the noise-suppressed band-limited image signal. The adder 44 obtains a sum of the processed band-limited image signal and the interpolated (magnified) image signal.

[0105]

FIG. 13 is a diagram schematically illustrating an operation for performing the Laplacian reconstruction. After the processed band-limited image signals fBk in which noise components are suppressed (k=1 to n) are obtained, the interpolation processing unit 43 performs interpolation processing, as with the aforementioned interpolation processing unit 11, on the processed band-limited image signal fBn at the lowest resolution level so as to obtain an interpolated magnified image signal Sn'. Then, the magnified noise component signal Sn' is added to the noise component signal fBk-1 by the adder 44 so as to produce an added noise component signal Sn-1 at the second lowest resolution level. Such processing is repeatedly performed so as to obtain a higher resolution, thereby providing an added noise component signal S1 with the highest resolution level. The added noise component signal S1 with the highest resolution level is processed so as to obtain a noise-suppressed image signal Sproc.

[0106]

Thus, when an image is output based on the processed image signal Sproc, "an image with a low exposure dose can obtained an image having an appearance which is similar to the appearances of slightly unsharped images of high-frequency components" by controlling parameters in the adaptive filter and the like. Since the weights in the weighted average of the neighborhood averages of vectors at a resolution level of interest and a lower resolution level are controlled based on the X-ray dose or doses in the input image, fine-edge-oriented noise suppression is performed in areas of an image which are exposed with a high X-ray dose, and large-edge-oriented noise suppression is performed in areas of the image which are exposed with a low X-ray dose. Therefore, noise can be effectively suppressed or eliminated in the entire image, edge degradation, which can be caused by noise suppression in a noisy image, can be reduced, and variations in image quality caused by variations in the exposure dose can be suppressed. In other

words, even when the amount of noise included in the image varies due to variations in the exposure dose, the noise suppressing apparatus can effectively suppress the noise in the image, reduce artifacts (unnaturalness) such as an arabesque pattern, and make the image more natural. In addition, degradation of fine signals can be reduced. Thus, it is possible to obtain high quality images.

[0107]

While the noise suppressing apparatus of the invention has been described in terms of preferred embodiments, it should be understood that the present invention not necessarily limited thereto.

[0108]

Although in the described embodiments, the band-limited image signals in respectively different frequency bands are generated from the input image signal Sin by the Laplacian pyramid decomposition, the band-limited image signals may be generated by the wavelet transformation as disclosed in Japanese Unexamined Patent Publication No. 6(1994)-274615.

[0109]

Although in the described embodiments, the noise-suppressed image signal is obtained from the plurality of noise-suppressed band-limited image signals by the inverse multiresolution transformation, an added noise component signal SH1 representing all of the noise components may be obtained by using the multiple resolution signals, and subtracted from the input image signal Sin so as to obtain the noise-suppressed image signal Sin, as disclosed in Japanese Patent Application No. 11(1999)-363766. Fig. 14 is a block diagram illustrating details of such a noise suppressing apparatus 100.

[0110]

In the described embodiments, vector information is used as information locally calculated from pixel values in the vicinity of a pixel of interest (hereinafter referred to as

local information), and the index values indicating the degree of noise suppression is obtained based on the vector information (the weighted average of neighborhood averages of the vectors), and filter processing is performed based on the index values. Alternatively, a moving average deviation may be used as the local information, instead of the above vector information. In the case where the moving average deviation is used, when a first moving average deviation corresponding to a first band-limited image signal is calculated, a second moving average deviation may be added to the first moving average deviation according to information which indicates an exposure dose such as a menu or condition of radiography or an amount corresponding to the radiation dose, where the second moving average deviation is calculated for the pixel of interest from a second band-limited image signal at a second resolution level which is lower than the first resolution level.

[0111]

In the smoothing processing, the filter coefficients or mask size of the (normal two-dimensional) isotropic spatial filter may be changed.

[0112]

The characteristics of the filter may be changed based on the local variance of the detail image and information corresponding to the X-ray dose, where the local variance of the detail image can be obtained, for example, by using a method disclosed in Japanese Unexamined Patent Publication No. 6(1994)-96200 mentioned above. However, compared with the smoothing processing along the orientation of the edge, edge degradation is likely to occur in the smoothing processing by changing the characteristics of the isotropic spatial filter. Therefore, the smoothing processing along the orientation of the edge is more preferable.

[0113]

In the described embodiments, noise properties of the

input image signal is calculated based on the information on the exposure dose after the plurality of band-limited image signals respectively representing band-limited images in different frequency bands are produced, and a characteristic of the smoothing filter are adapted based on the calculated noise properties so as to smooth each band-limited image signal by using the smoothing filter. Alternatively, even when smoothing processing is performed on the input image signal per se, it is possible to adaptively change characteristics of a smoothing filter based on the characteristics of the input image signal. In this case, it is also possible to use the aforementioned techniques in which the noise properties is obtained by using local information such as the vector information as well as the information on the exposure dose.

[0114]

In addition, the noise suppressing method described above may be carried out by an computer, and a program for instructing the computer to carry out the noise suppressing method may be stored in the computer-readable storage medium and such a computer-readable storage medium may be provided.

[Brief Description of the Drawings]

[FIG. 1]

FIG. 1 is a schematic block diagram illustrating a construction of a noise suppressing apparatus according to one embodiment of the invention.

[FIG. 2]

FIG. 2 is a flow diagram illustrating a processing sequence of the noise suppressing apparatus.

[FIG. 3]

FIG. 3 is a block diagram illustrating an outline of a construction of the band-limited-image-signal generation unit.

[FIG. 4]

FIG. 4 is a diagram schematically illustrating the operations of generating band-limited image signals.

[FIG. 5]

FIG. 5 is a block diagram illustrating details of a general configuration of the noise suppressing apparatus.

[FIG. 6]

FIG. 6 shows conceptual diagrams for illustrating the double-angle representation.

[FIG. 7]

FIG. 7 is a diagram illustrating the definition of the D-A representation.

[FIG. 8]

FIG. 8 shows views respectively illustrating four two-dimensional filters.

[FIG. 9]

FIG. 9 is a diagram illustrating a relationship between a Laplacian signal and an output of a first-derivative filter (the absolute value of the first derivative).

[FIG. 10]

FIG. 10 shows views respectively illustrating isotropic two-dimensional filters.

[FIG. 11]

FIG. 11 shows views respectively illustrating recognition models of a line, a point of intersection, an end point, and noise based on a degree of edge confidence and an index of pixel energy.

[FIG. 12]

FIG. 12 shows views respectively illustrating examples of anisotropic filters having various characteristics.

[FIG. 13]

FIG. 13 is a diagram schematically illustrating an operation for performing the Laplacian reconstruction.

[FIG. 14]

FIG. 14 is a detailed block diagram illustrating a construction of a noise suppressing apparatus according to another embodiment of the invention.

[Explanation of the Reference Numerals]

100 noise suppressing apparatus

1 band-limited-image-signal generation unit

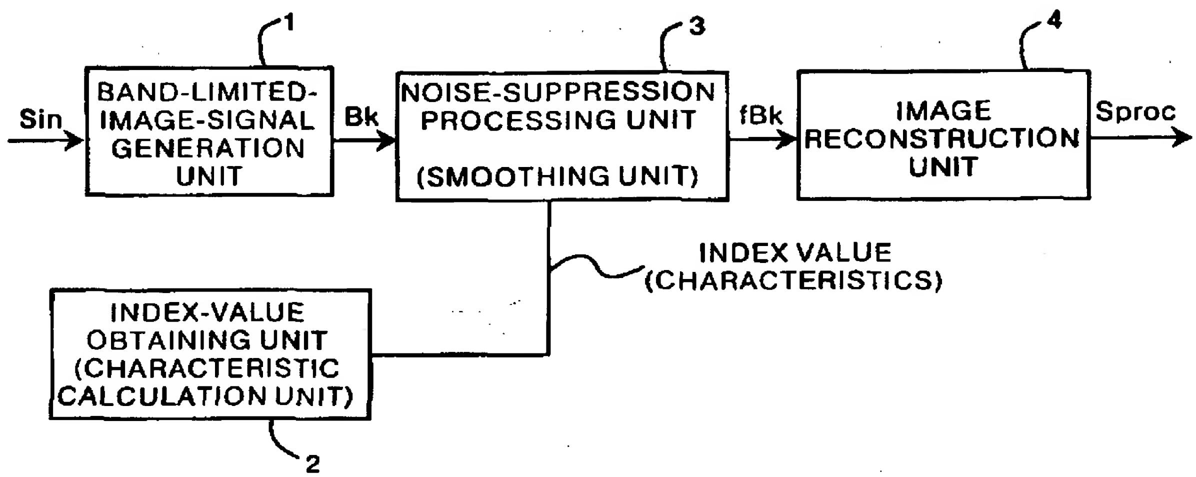
2 index-value obtaining unit (noise properties calculating means)

3 noise-suppression processing unit (smoothing means)

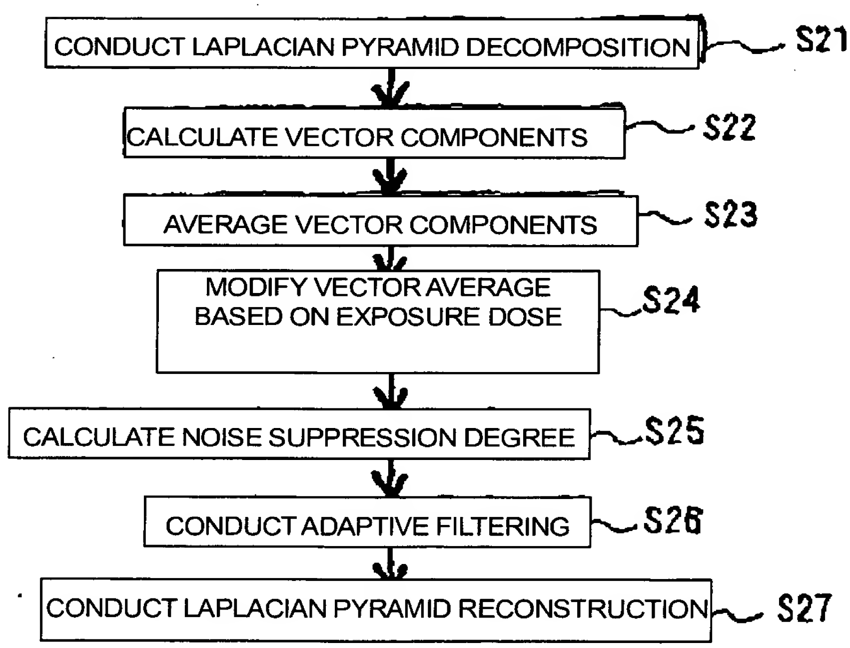
4 image reconstruction unit

[NAME OF DOCUMENT] DRAWINGS

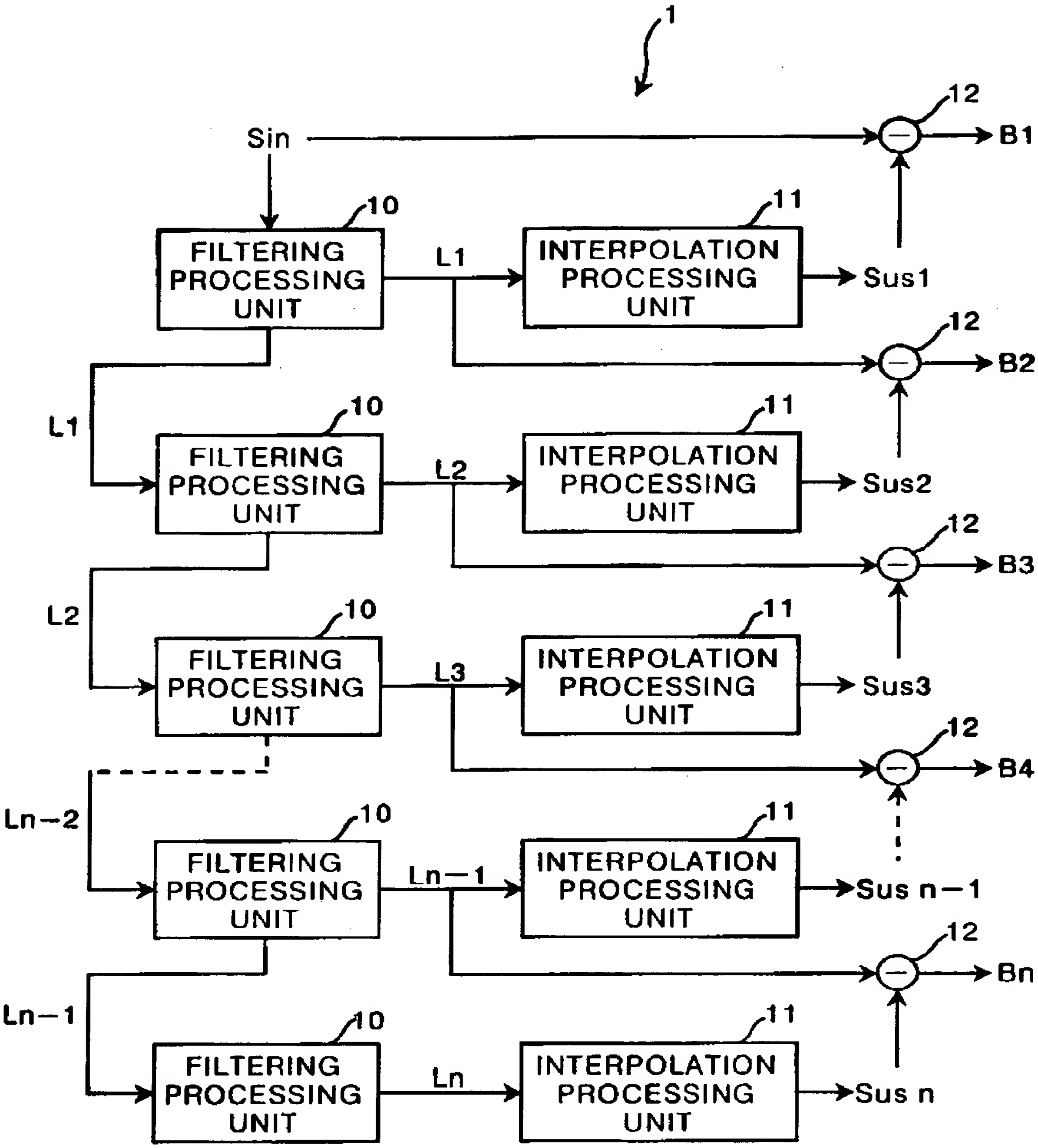
[FIG. 1]



[FIG. 2]

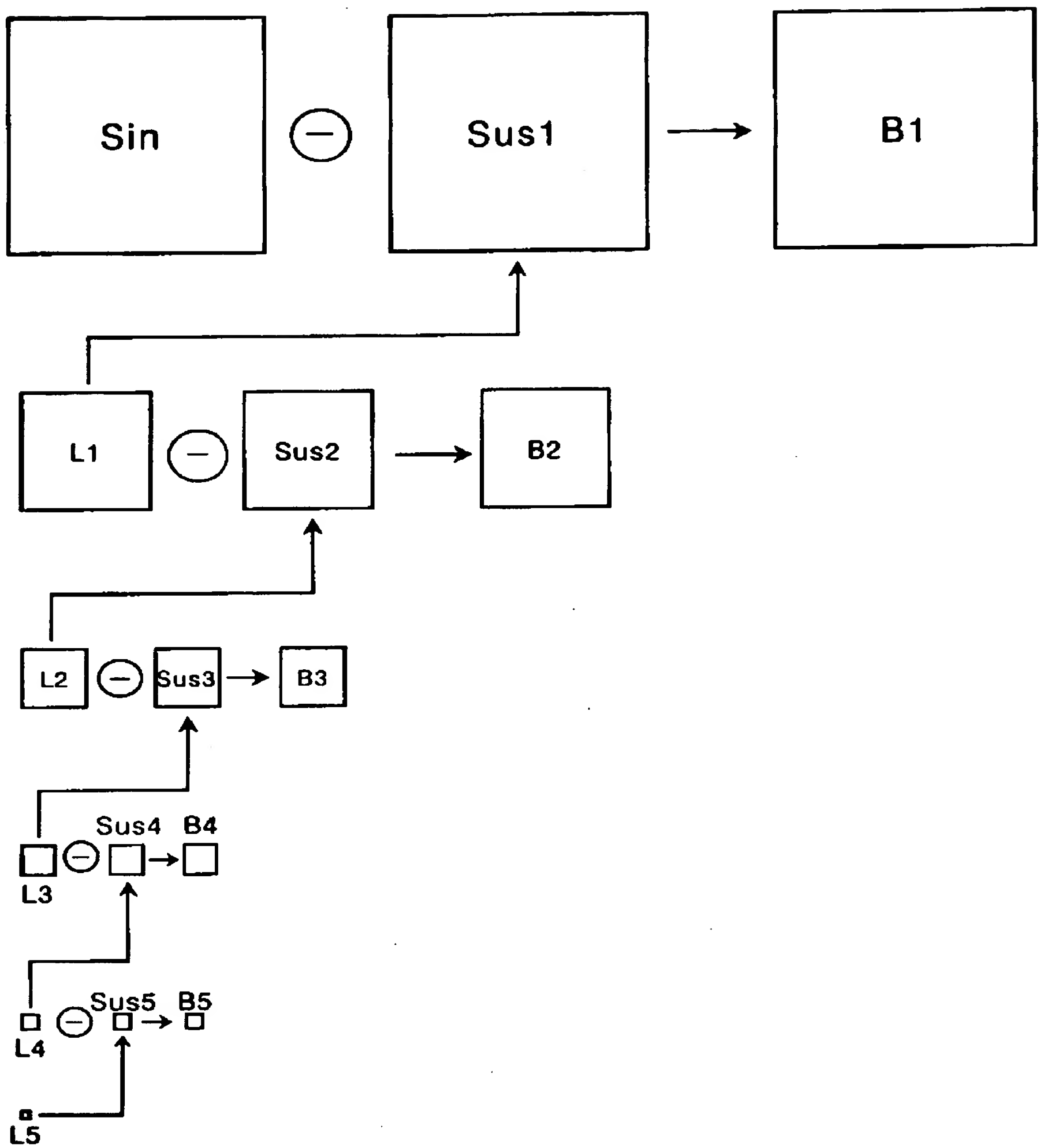


[FIG. 3]

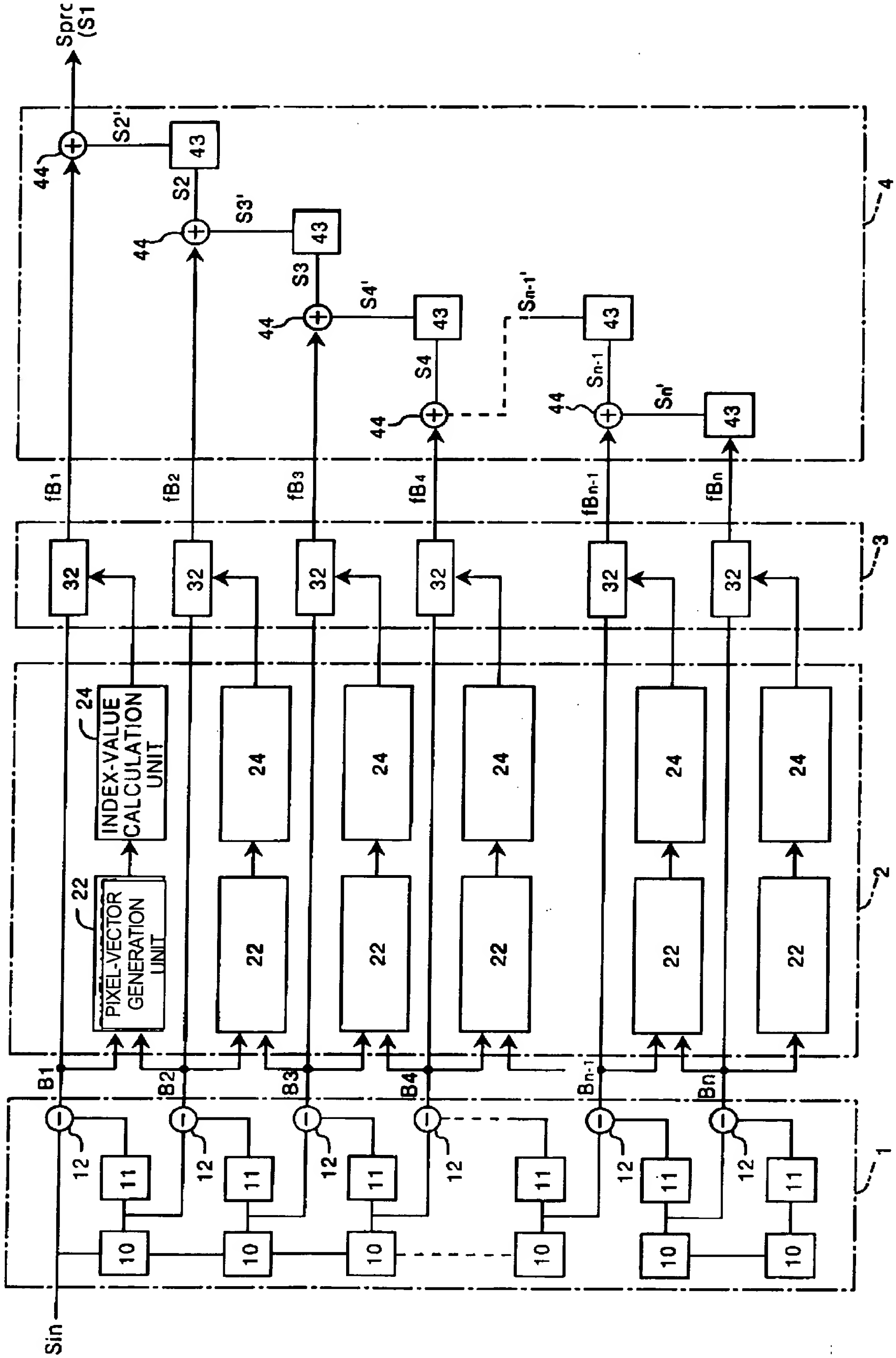


1
Sin
12
B1
10
11
FILTERING PROCESSING UNIT
L1
INTERPOLATION PROCESSING UNIT
Sus1
12
B2
L1
10
11
FILTERING PROCESSING UNIT
L2
INTERPOLATION PROCESSING UNIT
Sus2
12
B3
L2
10
11
FILTERING PROCESSING UNIT
L3
INTERPOLATION PROCESSING UNIT
Sus3
12
B4
Ln−2
10
11
FILTERING PROCESSING UNIT
Ln−1
INTERPOLATION PROCESSING UNIT
Sus n−1
12
Bn
Ln−1
10
11
FILTERING PROCESSING UNIT
Ln
INTERPOLATION PROCESSING UNIT
Sus n

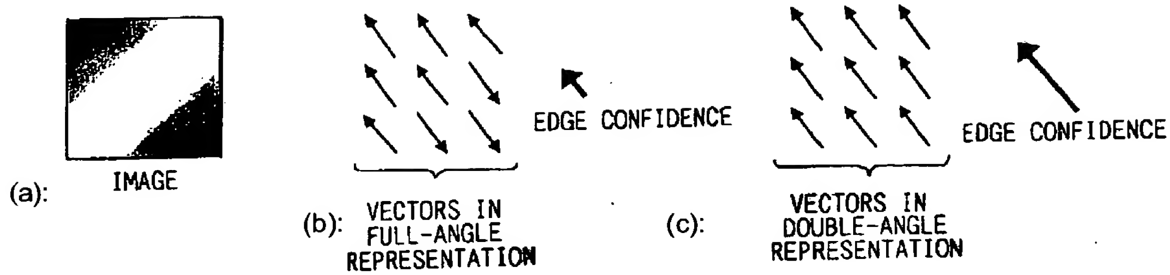
[FIG. 4]



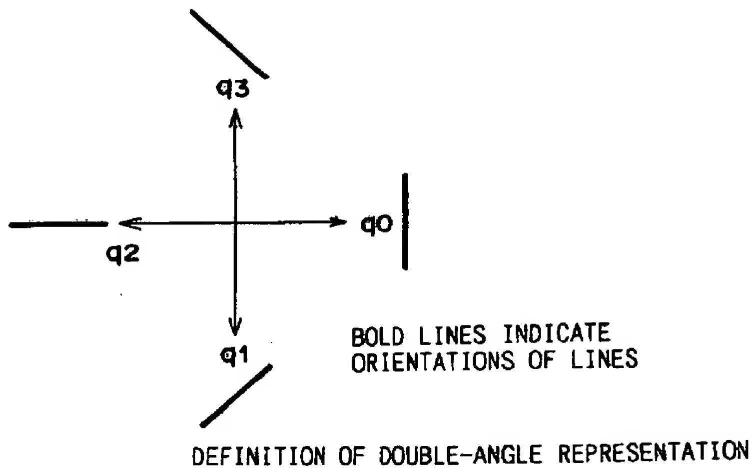
[FIG. 5]


Sin
Sprc (S1)
PIXEL-VECTOR GENERATION UNIT
INDEX-VALUE CALCULATION UNIT
10
11
12
B1
B2
B3
B4
Bn-1
Bn
22
24
32
fB1
fB2
fB3
fB4
fBn-1
fBn
43
44
S2
S2'
S3
S3'
S4
S4'
Sn-1
Sn-1'
Sn'
1
2
3
4

[FIG. 6]

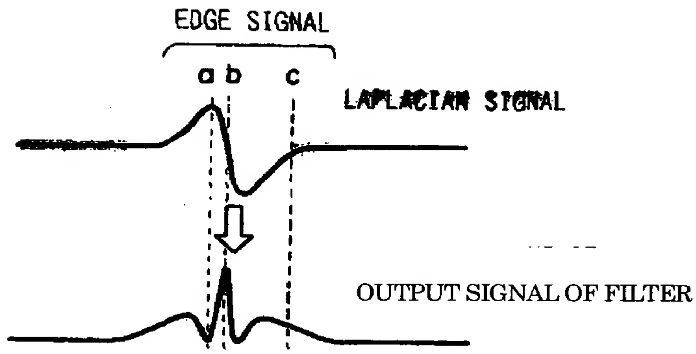


[FIG. 7]



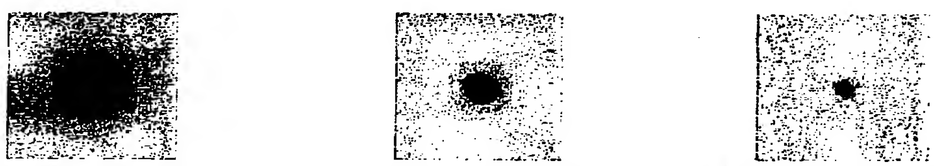
[FIG. 8]

[FIG. 9]



RELATIONSHIP BETWEEN LAPLACIAN SIGNAL AND OUTPUT OF FIRST-DERIVATIVE FILTER (ABSOLUTE VALUE OF FIRST DERIVATIVE)

[FIG. 10]

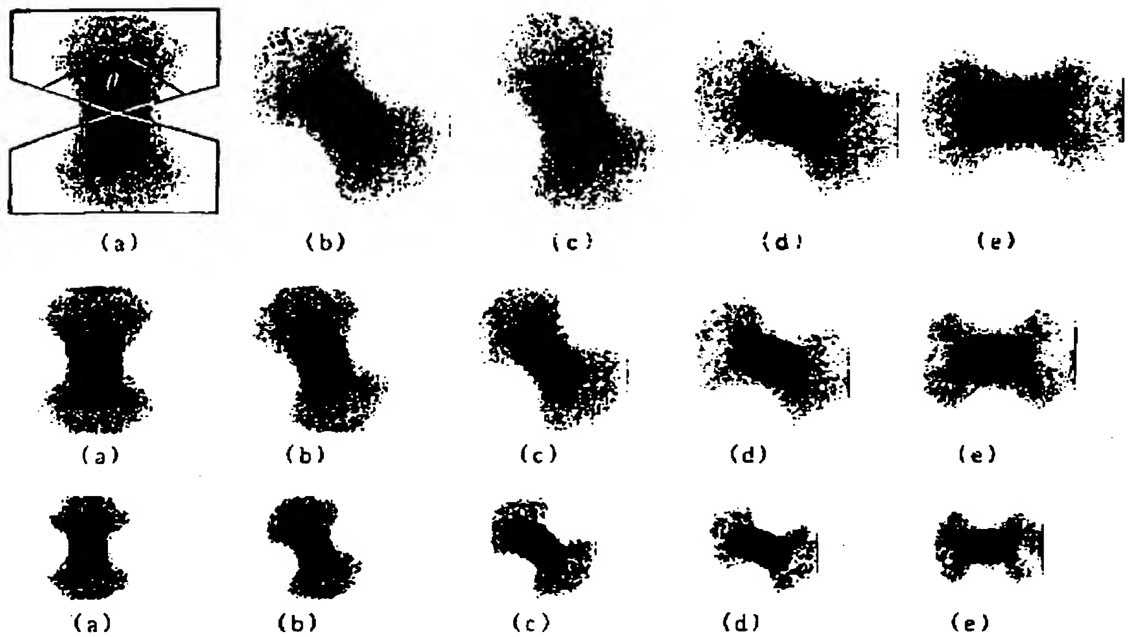


(a): FOR TARGET RESOLUTION LEVEL (b): FOR ONE-LEVEL LOWER RESOLUTION LEVEL (c): FOR TWO-LEVEL LOWER RESOLUTION LEVEL

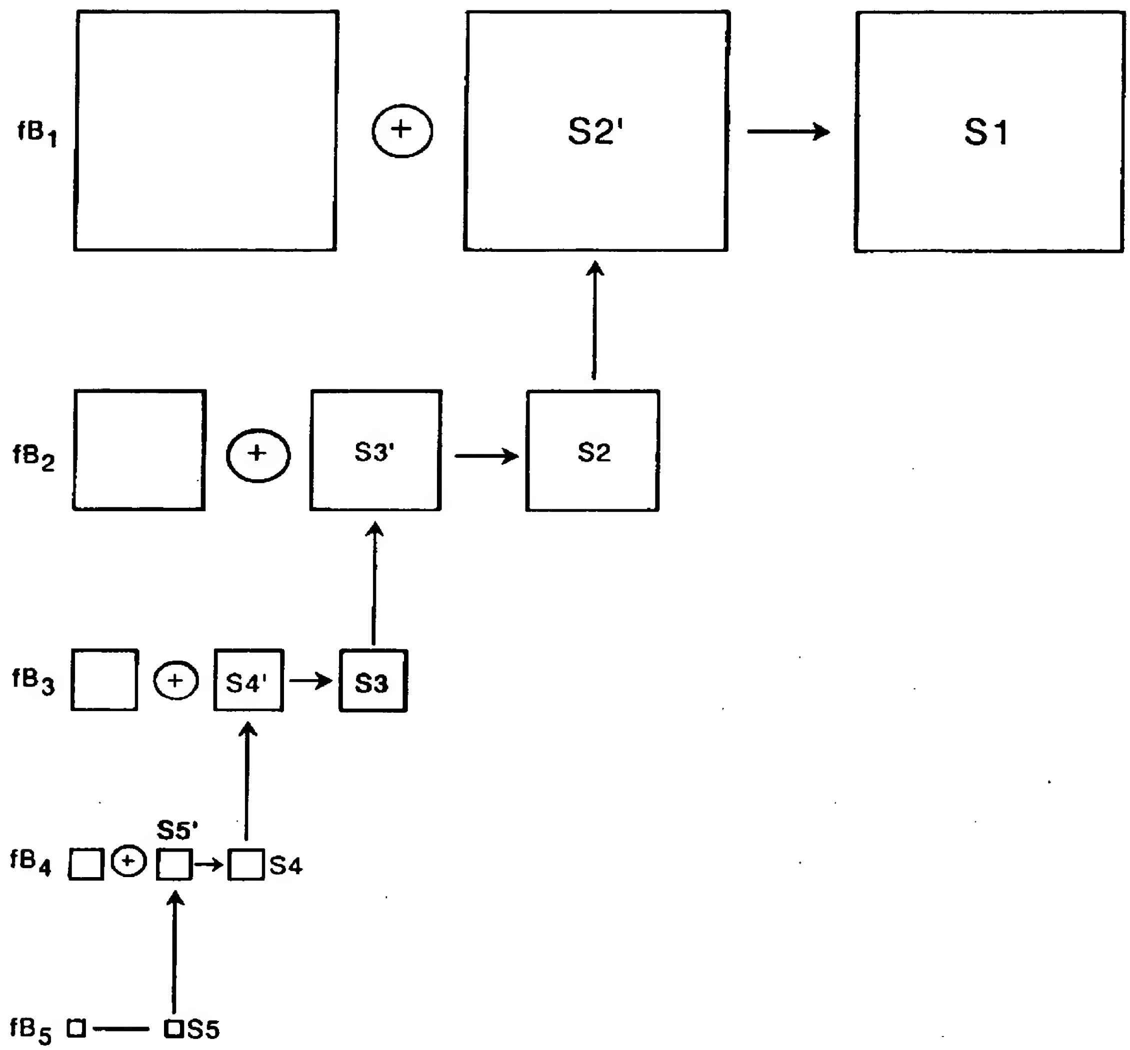
[FIG. 11]

(a): LINF $C \fallingdotseq 1$ & $E \fallingdotseq 1$ (b): LINE/NOISE $C \fallingdotseq 1$ & $E << 1$ (c): INTERSECTION POINT OR END POINT $C << 1$ & $E \fallingdotseq 1$ (d): NOISE $C << 1$ & $E << 1$

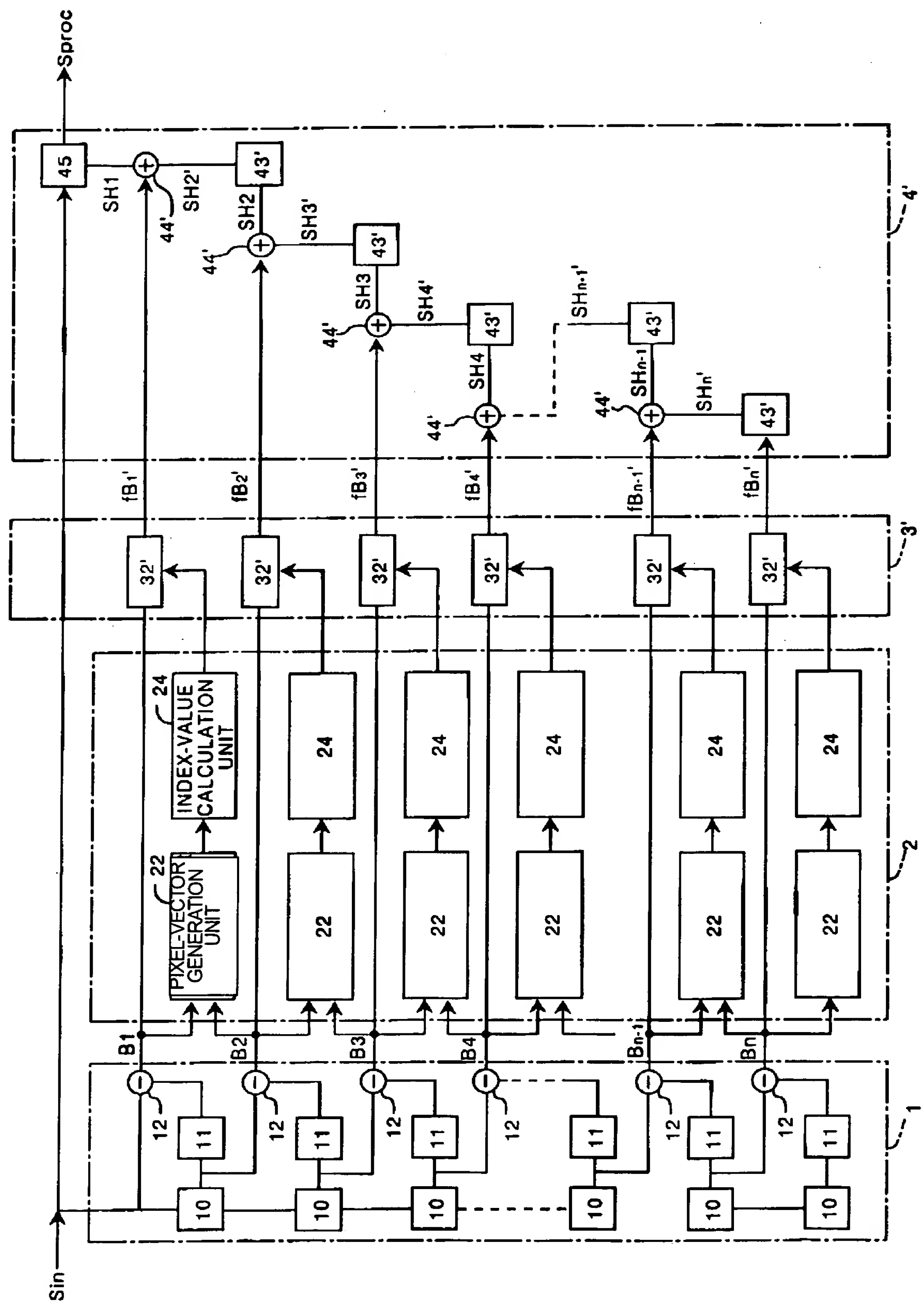
[FIG. 12]



[FIG. 13]



[FIG. 14]



Sin
Sproc
PIXEL-VECTOR GENERATION UNIT
INDEX-VALUE CALCULATION UNIT
B1
B2
B3
B4
Bn-1
Bn
fB1'
fB2'
fB3'
fB4'
fBn-1'
fBn'
SH1
SH2
SH2'
SH3
SH3'
SH4
SH4'
SHn-1
SHn-1'
SHn'
10
11
12
22
24
32'
43'
44'
45
1
2
3'
4'

[Name of Document] ABSTRACT

[Abstract]

[Objective] A noise suppressing apparatus for suppressing noise components included in a radiographic image regardless of an exposure dose is provided which can effectively suppress the noise regardless of an exposure dose.

[Constitution] A noise suppressing apparatus 100 has a band-limited-image-signal generation unit 1 for generating a plurality of band-limited image signals Bk that represent a plurality of band-limited images respectively belong to a plurality of different frequency bands; an index-value obtaining unit 2 for obtaining a pixel vector in a pixel of interest in a band-limited image and detecting an orientation of an edge as noise properties by using the pixel vector; a noise-suppression processing unit 3 for adapting characteristics of smoothing filters so that smoothing processing can be performed along the orientation of the detected edge and performing the smoothing processing on each of the plurality of band-limited image signals Bk by use of the smoothing filter after subjecting to the adaptation of characteristics; and an image reconstruction unit 4 for reconstructing a processed image signal Sproc which represents a noise-suppressed image, from the band-limited image signals fBk on which undergoes suppression of the noise components.

[Selected Figure] FIG. 1